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Real time target tracking in a phantom using ultrasonic imaging

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Abstract

In this paper we present a real-time ultrasound image guidance method suitable for tracking the motion of tumors. A 2D ultrasound based motion tracking system was evaluated. A robot was used to control the focused ultrasound and position it at the target that has been segmented from a real-time ultrasound video. Tracking accuracy and precision were investigated using a lesion mimicking phantom. Experiments have been conducted and results show sufficient efficiency of the image guidance algorithm. This work could be developed as the foundation for combining the real time ultrasound imaging tracking and MRI thermometry monitoring non-invasive surgery.

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Keywords: ultrasound image guiding, robot, real-time

1. Introduction

Focused ultrasound surgery (FUS) is developing rapidly as a completely non-invasive medical intervention alternative. Application of FUS in the treatment of fibroabdoma of uterus has passed the FDA clearance in 2004 and the sonication of bone metastasis has obtained a CE mark (Chen, 2005). Other tumours are under preclinical (prostate, kidney) or clinical (brain, breast, liver) evaluations.

However, the requirement for motion management during FUS treatment is recognized. Motion of the target object due to breathing, peristalsis, heartbeat or discomfort caused by FUS ablation is critical during FUS interventions. To prevent ablating surrounding normal tissue rather than target lesion, several approaches exist based on using physical constraints to minimize target's motion. However, it is not comfort for patients to hold their breath for too long (Dawson et al., 2001).

There are several treatment control methods which might solve this problem. The forms of treatment control include gating (Korreman et al., 2005) (the treatment is only applied during a certain range of the breathing cycle), triggering (the treatment is applied for a certain time period as soon as a certain phase of the breathing cycle has

been reached) or adaptive methods, where the beam follows the motion. The last method is favorable since the treatment is not interrupted and minimal duration of intervention is achieved with a maximum accuracy possible.

Typically MRI images provide best guidance to focused ultrasound surgery because of its good soft tissue contrast, temperature monitoring (Yea et al., 2005) and tissue coagulation detection. However, MR image acquisition time is slow, therefore it is not preferable for real-time monitoring when the target object is moving (Bock et al., 2004). Ultrasound, with real-time imaging, more cost effective than MRI, is a promising alternative.

In this paper, the feasibility of ultrasound-guided tracking was investigated in order to compensate for involuntary patient movement. Ultrasound scanner was used to image the target as real time visual servoing to guide a robot's motion. The real time ultrasound image tracking guided focused ultrasound ablation control was realized.

The ultrasound tracking system consists of an ultrasound imaging probe which is 7.5MHz, a 6-DOF industrial robot, a single-element arch FUS transducer (1MHz), and a computer workstation with software to process video. The system layout is illustrated in Figure 1.

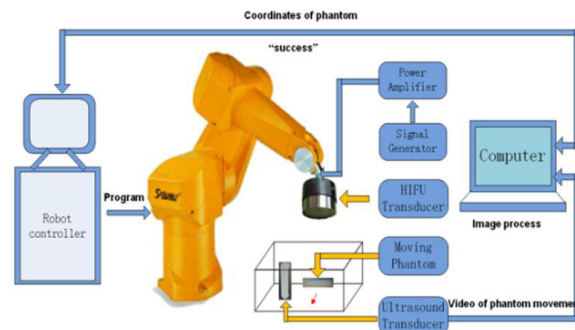


Figure 1. Imaging and tracking system, red arrow points the movement direction of the tumor phantom.

The ultrasound probe acquires the continuous images of the tumor phantom in motion. Calculated coordinates from the image processing workstation are transferred to the robot control workstation which will guide the FUS transducer moving along with the phantom in real-time.

The working procedure of the software is shown in Figure 2. Selected image processing algorithms have been developed here to extract the contour of the target, and identify the moving phantom section of the target simultaneously in real time. Phantom with known geometry is used to verify the system.

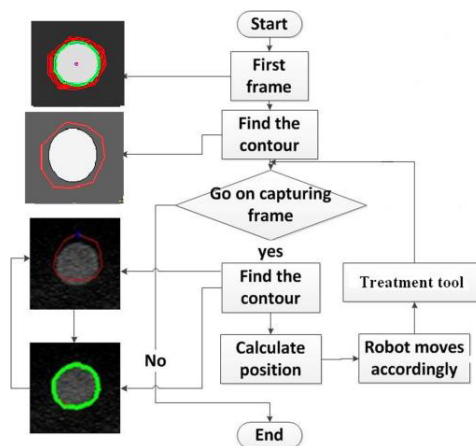


Figure 2. The whole programme procedure

Active snakes based on the gradient vector flow has preliminary chosen for the tracking software. Several key issues have been addressed in contour extracting and real-time tracking. Firstly, the algorithm using active snake contour was applied to find targets with various irregular shapes. It locks onto nearby edges, localizes the contours accurately. Using the gradient vector flow (GVF) method proposed by Xu in 1998, the snake can converge to boundary concavities and not leak from the discontinuous boundary. Secondly, the snake makes good use of the relationship between every two successive frames. When the snake is computing the boundary at the current frame, it can take the result boundary from the last frame as the initialization breed for the current one. Thirdly, as a recursive algorithm, the active snake contour algorithm needs a good computing speed. Several methods have been developed to reduce the computing load of active snakes (Han et al., 2007). By utilizing these methods properly, the computing speed can be increased to a certain extent.

Active snake contour is an energy minimizing deformable curve which first introduced by Kass etc. A few sample points are required to describe the boundary of the target object by $\mathbf{v}(s) = (x(s), y(s))$. The energy function is written as

$$\begin{aligned} E_{snake}^* &= \int E_{snake}(\mathbf{v}(s)) ds \\ E_{snake}^* &= \int E_{int}(\mathbf{v}(s)) + E_{ext}(\mathbf{v}(s)) ds \\ E_{snake}^* &= \int \frac{1}{2} [\alpha |X'(s)|^2 + \beta |X''(s)|^2] + E_{ext}(\mathbf{v}(s)) ds \end{aligned}$$

where E_{int} is the internal force of the spline which acts as a smooth constraint to the contour shape, E_{ext} is the external force which drives the spline towards desired image features, α and β represents elasticity and rigidity of the contour respectively. The external force function E_{ext} is derived from the image, it takes on its smaller values at the features of interest, usually boundaries.

The gradient vector flow (Xu, 1998) is used for the tracking here due to that it is insensitive to the contour initialization and is able to deform into concave part of the object unlike conventional contour models.

Typically, the required number of iterations of the external forces for an image of $a \times b$ pixels is $n = \sqrt{a \times b}$. When the GVF is applied on large images, the computing load will increase dramatically. For the images used in the experiment, the size is around 720×576 pixels, the computing time for the external forces using GVF is over 25s, which is not acceptable for real time application.

SUGVF (speed up GVF) and MGVF (multi GVF) are among the most efficient methods to speed up the computing of GVF. However, the SUGVF doesn't focus on medical images and doesn't present enough details on how to incorporate the SUGVF into the B-spline convergence mechanism. The MGVF is an alternative (Xu, 1998) with a multi-grid method, partly addressing the limitation of the original computing method.

By employing the multigrid GVF method, the speed up performance is shown in Table 1. For the image size 720×576 pixels, the computing speed can be decreased by 42 times. The performance is not as good as claimed by Xu, this is might because the noise conditions or the computing hardware is different. The computing efficiency for the external forces is increased dramatically.

When the internal forces E_{int} is taken into consideration, the total time consuming for an image of 720×576 is over 2 seconds, which is not acceptable for real time tracking applications. Modifications have been made to decrease the computing time further. Because the computing load is highly relevant to the image size and the number of iteration, carefully choosing image size and number of iteration can reduce the algorithm load. Because displacement of the targets between two successive frames is usually smaller than 4 or 5 pixels, a small Region of Interest (ROI) window is selected around the current target position to reduce the size of image processed. This ROI window is used to cover the next position of the target. Using this improved method, processing time needed for one single frame is short than 0.2s, this reduces around 10 times the amount of computing compared with original MGVF method, as shown in Table 2.

Table 1. Test for the performance of GVF and MGVF

Image size	Original GVF iterations	$\ \rho\ _m$	Time (s)	MGVF iterations	$\ \rho\ _{max}$	Time(s)	Speed gain
720×576	320	5.0×10^{-4}	63.20	2	4.8×10^{-5}	1.5	42

ρ : residual of two successive recursive computing, reflecting computing precision

Table 2. Speed gain after applying window on the image

Original image size	MGVF iterations	$\ \rho\ _m$	E _{ext} time (s)	Total time (s)	Window size	Total Time (s)	Speed gain
720×576	2	4.8×10^{-5}	1.5	>2.0	(70~80)×(70~80)	<0.2	10

The ultrasound probe submerges in a water tank of 50×50×100 cm cube while the tumor phantom moved horizontally. The ultrasound image plane is set perpendicularly to the axis of the phantom. A step motor controls the motion of the tumor phantom. The time cycle of moving is around 5.0s and the amplitude is around 20mm which mimics the parameters of liver under quiet breathing motion (Davies et al., 1994). The cylinder shaped phantom is made of homogeneous egg white and polyacrylamide (PAA) which is a kind of transparent gel. This composition of materials will be easily and rapidly ablated by FUS (Takegami et al., 2004).

A single element ball-shaped FUS transducer is used. The transducer was working under the power of about 30 Watts, supplied by a RF power amplifier.

FUS experiments have been setup as shown in

Figure 3. The robot guides the FUS transducer to deliver ablation to the moving phantom.

The movement accuracy testing was conducted. By comparing the displacement of the phantom and the robot, the performance of the algorithm was satisfied with a standard deviation of the errors ($\pm 0.18\text{mm}$), which is small compared to the size of target (20mm). System error in real time was also estimated, which includes time delays caused by the image processing time and robot, the total system error is around 0.39mm.

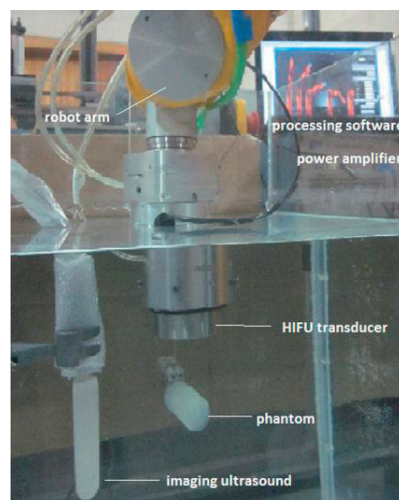


Figure 3 Experiment setup. The ultrasound imaging probe and the focal point of the FUS transducer are in a same plane which crosses the moving phantom

Any asynchrony caused by time delay in ultrasound guided FUS could result in either an insufficient dose of FUS in the malignant tissue, or an overdose in the surrounding normal tissue mistakenly ablation surrounding normal

tissue. In this paper, time delay at every step was measured, the total time delay is around 0.2s including image acquisition, image process, and data transferring between devices.

There are two other causes contributing to random measurement errors. First is the image resolution, our imaging system is using one image pixel to indicate an area of 0.4×0.4 mm square after our calibration. This might become an accuracy limitation when tracking smaller lesions. The second factor is also an algorithm error when computing the centre of the object. One pixel mistake could cause a 0.4mm positioning error as mentioned.

The noise cause by focused ultrasound should also be considered. Pulsed focused ultrasound can replace the continuous one. The image acquisition and processing can be conducted at the interval between two focused ultrasound durations, as shown in Figure 4a. The image is much clear when the pulsed focused ultrasound is used as shown in Figure 4c. The noise straps can be removed during image processing. As an alternative, the image can be captured only when the pulsed ultrasound is switched off by triggering method.

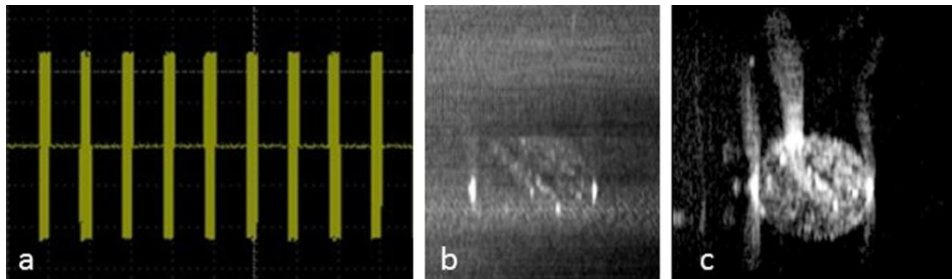


Figure 4(a) Pulsed focused ultrasound; (b) Image captured when continuous FUS is applied; (c) Image captured when pulsed FUS is applied.

2. Conclusion

In this study, an ultrasound image real time guided FUS ablation system has been setup including imaging device, robot and a moving phantom. By utilizing active snake algorithm, a tracking programme was compiled with a high speed (5Hz) acceptable for real time target extracting.

With a standard ultrasound imaging machine, this tracking technique offers an accuracy of ± 0.18 mm in position and position error duo to time delay in tracking has also estimated below 0.39mm.

We believe that this tracking system offers a possible solution for tracking moving tumor inside human liver for invasive focused ultrasound surgery.

Acknowledgement

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